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Surgical performance when inserting non-locking screws: a systematic review

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- Billions of screws are inserted by surgeons each year, making them the most commonly inserted implant. When using non-locking screws, insertion technique is decided by the surgeon, including how much to tighten each screw. The aims of this study were to assess, through a systematic review, the screw tightness and rate of material stripping produced by surgeons and the effect of different variables related to screw insertion.
- Twelve studies were included, with 260 surgeons inserting a total of 2793 screws; an average of 11 screws each, although only 1510 screws have been inserted by 145 surgeons where tightness was measured – average tightness was 78±10% for cortical ($n = 1079$) and 80±6% for cancellous screw insertions ($n = 431$).
- An average of 26% of all inserted screws irreparably damaged and stripped screw holes, reducing the construct pullout strength. Furthermore, awareness of bone stripping is very poor, meaning that screws must be considerably overtightened before a surgeon will typically detect it.
- Variation between individual surgeons' ability to optimally insert screws was seen, with some surgeons stripping more than 90% of samples and others hardly any. Contradictory findings were seen for the relationship between the tightness achieved and bone density.
- The optimum tightness for screws remains unknown, thus subjectively chosen screw tightness, which varies greatly, remains without an established target to generate the best possible construct for any given situation. Work is needed to establish these targets, and to develop methods to accurately and repeatably achieve them.

Keywords: bone screws; fracture fixation; internal fixation; screw insertion; screw tightness; stripping torque; surgical technique

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Background

The quality and efficacy of orthopaedic fixation relies on screw design and material, bone characteristics and surgical techniques. Traditional fixation methods using non-locking screws, to generate compression and stability, remain important despite an increased use of locking screw constructs.[1] When inserting non-locking screws, friction is generated between screw threads and the host bone to produce a shear force and counteract the linear motion of the screw. This friction enables stabilization and compression of bones and their fragments during locomotion to resist muscle and joint forces.

For non-locking screws, the force applied for tightening is subjectively chosen and controlled by the surgeon. If the torsional force applied to a screw exceeds the shear limit of the surrounding bone, the screw 'strips' the bone, reducing the resistance to pullout force by more than 80%.[2,3] This is an irreversible situation due to plastic deformation of the bone. These weakened constructs increase the risk of fixation failure, which doubles treatment costs and worsens patient morbidity and mortality.[4]

Attempts to quantitatively and qualitatively describe surgeons' abilities to insert screws have been performed; here we systematically review the existing work in this field. The first aim was to report the tightness of an inserted screw when expressed as a ratio (the stopping/stripping torque ratio) against the minimum stripping torque, where the stripping torque represents the upper limit of the tightening torque needed to strip the surrounding bone. The second aim was to identify the percentage of screws that are inserted beyond the stripping limit of the bone (beyond the stripping torque). The surrounding material is described as 'stripped' when the torque applied during insertion exceeded the maximum that can be resisted by the bone of the screw hole, causing it to yield. The third aim was to assess the association between surgical experience and stripping rates for the test material. The fourth aim was to assess the effect of different instructions given to surgeons on screw tightness and material stripping rates. Finally, the fifth aim was to study the effect of variations in bone density on screw tightness and material stripping rates.

Methods

Due to the nature of the data presented, a systematic review was performed in line with the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) guidance.[5] The search strategy employed free and Medical Subject Headings (MeSH) search terms and a combination of keywords relating to qualitative ("screw tightness", "overtightened", "tightness perception", "screw insertion") and quantitative screw insertion ("insertion torque", "stopping torque", "stripping torque", "stopping/stripping ratio"). There were no restrictions on publication dates. MEDLINE, EMBASE, Web of Science and the Cochrane Library electronic databases were searched up to 31 August 2018. Only articles in English and German were considered. Initial screening was performed in English by the lead author and in German by the second author using translations of the same keywords. Studies with any number of participants and any number of screw insertions were included. All bone models were included – human (in vivo and cadaveric), animal and artificial. For studies to be included for review of screw tightness, stopping and stripping torque values needed to be reported in order to calculate the tightness as a percentage, if this had not been calculated within the studies themselves. Exclusion criteria were failure to provide results for screw tightness and/or stripping rates for manually inserted non-locking screws. The reference lists of the included articles were manually scanned for any relevant additional studies. Calculated percentages are presented as integers.

Results

Our literature searches identified 2158 potentially relevant studies (Fig. 1). A further review of the titles and abstracts reduced the number of potentially relevant studies to 30. On full reading, 18 were excluded as screws were not inserted by hand with the tightness recorded. This process was repeated in German but yielded no further articles for inclusion. The 12 remaining articles were included in the review for assessment of screw tightness, of which nine reported the stripping rates explicitly in their manuscripts.[6–17]

Screw tightness achieved as a percentage of the maximum

Several experimental studies have investigated the torque achieved by a surgeon (stopping torque) and compared this to the maximum possible torque. The maximum torque being determined at a separate time interval by using a torque meter to further tighten the screw until the maximum torque value is reached when the material is stripped (stripping torque). By defining the maximum tightness as 100% (stopping torque = stripping torque), the ratio of stopping torque to stripping torque enables presentation of the tightness for that insertion as a percentage of the maximum. Many variables, such as the type of screw used and the material they were inserted into, have been assessed, generating a range of different screw tightnesses achieved by surgeons (Fig. 2). The first major work on this topic was published by Cordey et al in 1980.[6] Sixty-three orthopaedic and general surgeons manually tapped and inserted one 4.5 mm cortical, stainless steel screw unicortically into one human cadaveric femur, aiming to apply 'optimal torque for a good fixation'. This procedure was repeated with 35 surgeons inserting the same screws into one human cadaveric tibia. Screws were tightened to $84\pm13\%$ (mean±standard deviation) and $88\pm18\%$, respectively. The authors found that 10 out of 108 screws (9%) were inserted beyond the stripping torque; it was not recorded whether this was detected by the surgeons. In the second part of their study, 36 surgeons were asked to insert three screws into human cadaveric tibiae using three different methods. First, they assessed the effect of different drilling techniques by using either a large air drill for making pilot holes whilst having radiographs available and being able to see the bone, and second in separate holes repeating the first method but with a small air drill instead. Finally, they asked for screws to be inserted with neither radiographs available nor sight of the bone; though methods for blinding surgeons were not stated. None of these experimental setups generated significant differences in screw tightness.

In 1995 McGuire et al asked 105 orthopaedic surgeons of various experience to insert three titanium and three

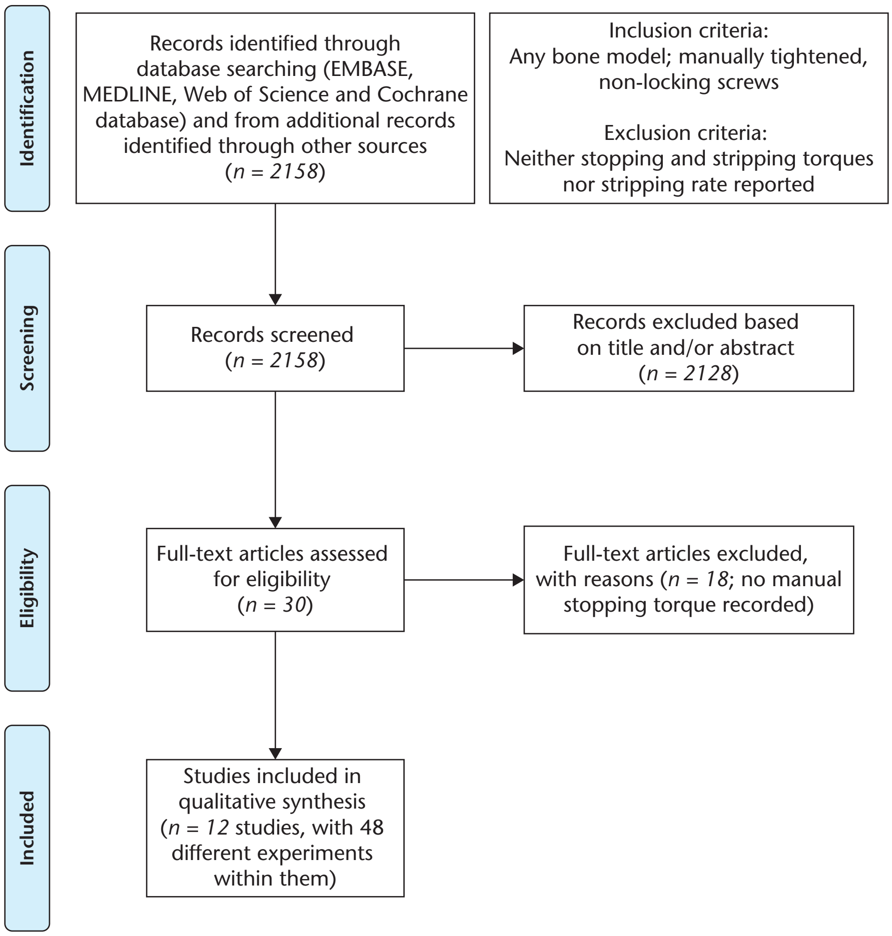

Fig. 1 PRISMA flow chart.

stainless steel 3.5 mm screws into non-locking plates.[7] The instructions to the surgeons were to insert the screws to what they considered 'two fingers tight using their normal technique'. This instruction being a subjective insertion method thought to reduce applicable torque, as a reduced grip is used due to only two fingers holding the screwdriver handle. The stopping torque was measured; however, no assessment of the stripping torque was performed. They found a significant trend for higher stopping torques with more years of surgical experience, a variable that will be explored later in this review. When inserting stainless steel screws, more torque was applied compared to titanium screws. Whilst the number of surgeons employed, and the number of screws inserted (n = 315 per screw type, three for each of 105 surgeons) is the largest of any study to date, this work was limited by the lack of stripping torque assessment, both whether any of the screws were stripped on insertion and whether or not this was detected.

Dinah et al had one surgeon inserting 200 screws (160 bicortical, 3.5 mm cortical screws and 40 unicortical, 4.0 mm cancellous screws) into human cadaveric fibulae.[14] They found that, on average, screws were inserted to 71% of the stripping torque. Analysis of their provided data actually shows this value to be 66% given that 83% of the inserted screws were cortical; the stopping/stripping torque ratio was 64% for cortical screws and 77% for cancellous screws. The range reported was 18% to > 100%, with values > 100% being calculated as the stopping torque was greater than the stripping torque that could be generated subsequently, as the material had already been stripped during the initial tightening episode.

Tsuji et al investigated the effect of bone density on tightness, in both human and artificial bone.[11] They measured average tightness for 24 insertions of 3.5 mm cortical screws in artificial bone of eight different densities, 12 insertions of 6.5 mm cancellous screws in Sawbones blocks of seven different densities, three insertions for 3.5 mm cortical screws into each of 16 human cadaveric femurs and two insertions of 6.5 mm cancellous screws into each of 16 cadaveric femoral condyles. Combining all densities, the tightnesses for cortical and cancellous

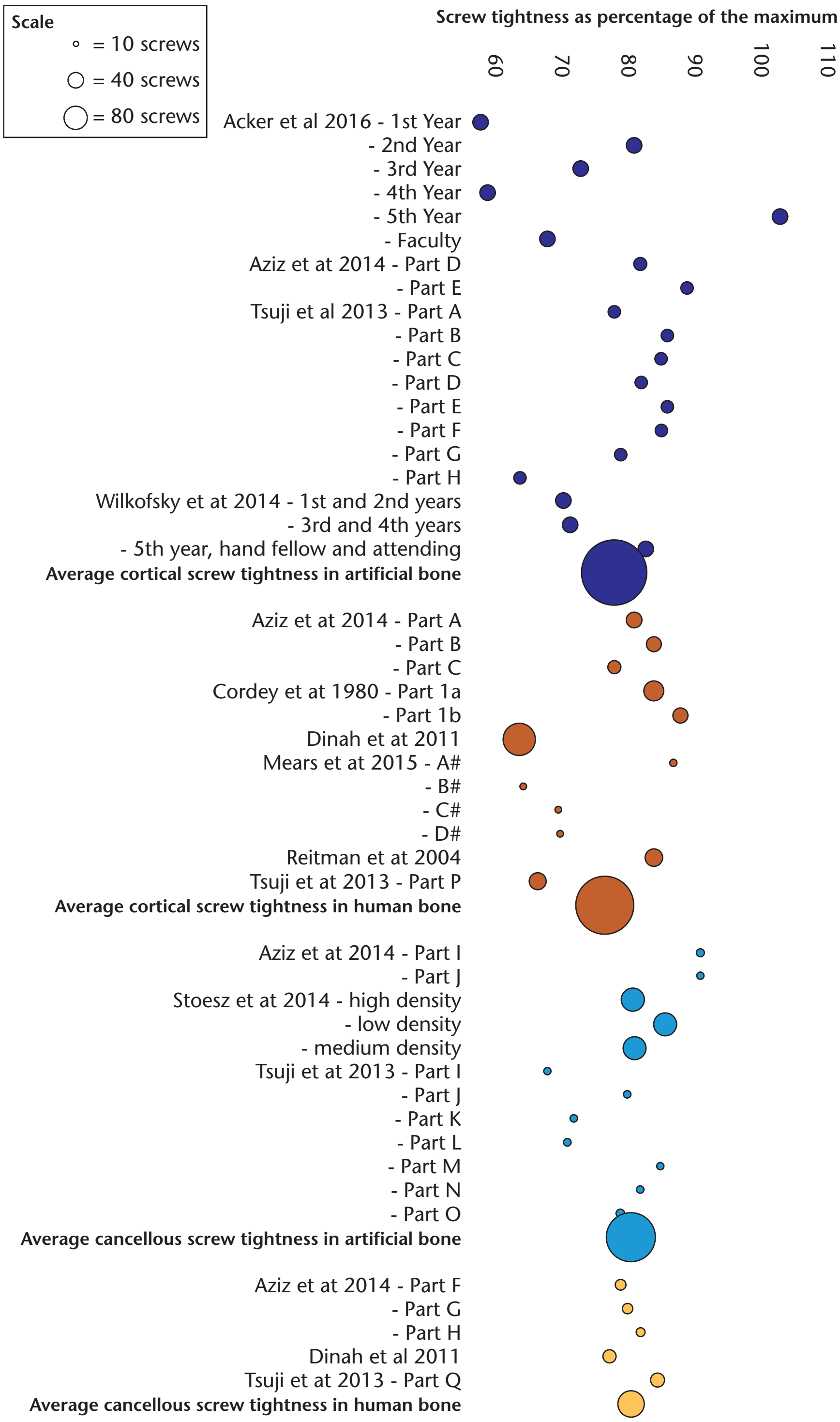

Fig. 2 Tightness achieved for each part within each study, where measured. From top to bottom, grouped alphabetically within the following sections: cortical screws in artificial bone (dark blue), cortical screws in human bone (dark orange), cancellous screws in artificial bone (light blue), cancellous screws in human bone (light orange). All bubbles scaled with size representing number of screws used, e.g. Acker et al 2016 – first-year = 40 screws. The different components of each study, where relevant, are explained as follows: Acker et al and Wilkofsky et al – different years of experience of surgeons; Aziz et al: A – cortical screws in fresh frozen human bone, B – cortical screws in embalmed human bone, C – cortical screws in dried human bone, D – cortical screws in normal density artificial bone, E – cortical screws in osteoporotic density artificial bone, F – cancellous screws in fresh frozen human bone, G – cancellous screws in embalmed human bone, H – cancellous screws in dried human bone; Tsuji et al artificial bone: densities for each part (cortical and cancellous screws respectively) – 0.08 g/cm^3 (A and I), 0.16 g/cm^3 (B and J), 0.24 g/cm^3 (C and K), 0.32 g/cm^3 (D and L), 0.40 g/cm^3 (E and M), 0.48 g/cm^3 (F and N), 0.64 g/cm^3 (G and O), 0.80 g/cm^3 (H, cortical only); Tsuji et al, human bone: P – cortical screws, Q – cancellous screws; Cordey et al, 1a – 4.5 mm cortical screws in human femur, 1b – 4.5 mm cortical screws in human tibia; Mears et al, A – 90° past contact of the screw head on the plate; B – 180° past contact of the screw head on the plate; C – two-fingers tight; D – 1.4 Nm; Stoesz et al, high density (0.32 g/cm3), medium density (0.16 g/cm^3), low density (0.08 g/cm^3). #Ratio estimated based on provided data, though not explicitly stated by authors.

Table 1. In vitro and in vivo percentages of bone samples stripped, the number of screws used within each study, the number of surgeons involved in descending screw number with methods described. When different variables tested or conditions changed within the same study, results have been separated into different 'parts' indicated with Roman numerals

Study	Percentage of bone samples stripped (%)	Number of screws inserted	Number of surgeons involved	Methods used
In vitro				
Stoesz et al, 2014[10]	45	240	10	4.0 mm cancellous screws in artificial bone (combined stripping rate for three densities as individual rates not provided)
Dinah et al, 2011[14]	4 (i)	160	1	Screws in human fibulae: (i) 3.5 mm cortical inserted bicortically; (ii) 4.0 mm cancellous screws inserted unicortically
	28 (ii)	40		
Cordey et al, 1980[6]	9 (i)	108	36	Cortical screws in human tibiae: (i) one screw inserted per surgeon under three conditions; (ii) one screw per surgeon inserted into three different bone densities
	2 (ii)	90	30 (of the previous 36 used in Part i)	
Gustafson et al, 2016[12]	42 (i)	80	10	4.0 mm cancellous screws in artificial bone: (i) baseline; (ii) with visual feedback; (iii) after visual feedback removed
	15 (ii)	80		
	35 (iii)	80		
Acker et al, 2016[9]	12 (i)	40	41	3.5 mm cortical screws in artificial bone. (i) first year, (ii) second year, (iii) third year, (iv) fourth year, (v) fifth year, (vi) faculty
	31 (ii)	40		
	24 (iii)	40		
	20 (iv)	40		
	53 (v)	40		
	19 (vi)	48		
Reitman et al, 2004[13]	2	48	1	3.5 mm cortical screws in human vertebrae bodies
Mears et al, 2015[15]	0 (i)	10	1	Cortical screws in human humeri: (i) 90° past contact of the screw head on the plate; (ii) 180° past contact of the screw head on the plate; (iii) two-fingers tight; (iv) 1.4 Nm
	30 (ii)	10		
	30 (iii)	10		
	20 (iv)	10		
In vivo				
Andreassen et al, 2004[17]	38	225	2	3.5 mm cortical and 4.0 mm cancellous screws in human fibulae
	Average reported stripping rate 26%	Total number of screws 1439	Total number of surgeons 102	

screws in artificial bone were 81% and 77% respectively, and in human cadaveric bone, 67% and 85% respectively. They did not report on the percentage of screws that stripped on insertion either quantitatively or subjectively. However, as average ratios were shown, at times averaging 24 tests, and that some tightness averages were 100%, it is likely that some screws stripped the samples on insertion.

Stoesz et al asked five senior resident and five senior practicing surgeons to each insert eight 4.0 mm cancellous screws into artificial bone of three densities (0.08, 0.16 and 0.32 g/cm^3).[10] Of the 239 screws reported, 131 were successfully inserted without stripping the bone; these had a tightness of 82±16%. The remaining screws (108 out of 239) stripped the polyurethane models. They found only a weak correlation (R^2 = 0.54) between surgeons who were able to insert screws close to, but below, the maximum and those who infrequently stripped screws. They also found that as surgeons inserted more screws in each density, stripping rates increased (p = 0.022); however, in another article employing similar methods with eight screws inserted, this effect was not seen.[12]

Acker et al asked 33 trainees and eight senior surgeons to insert six screws into bone models with a density of 0.48 g/cm^3, with the instructions to insert to 'two-finger tightness' with their dominant hand.[9] This was repeated with their non-dominant hand. Dominant hand data showed no significant difference between screw tightness when combining all surgical trainees (74%) and comparing this with faculty (68%); non-dominant hand data were not reported. The variability between participants grouped by years of experience, however, was large with first-year trainees' average tightness being 58% and fifth-years' 103%, i.e. the average for this latter group being beyond the stripping limit of the artificial bone. Additionally, there were large variations in achieved tightness within each group. This is the only study to have investigated the effect of hand dominance, finding a 70% difference in tightness between hands for first-year surgeons and 9% for senior surgeons.

Reitman et al asked one surgeon to insert screws into the anterior aspects of cadaveric human vertebral bodies.[13] Initially, one screw was inserted until stripped to establish the maximum torque, followed by a second screw into the same vertebral body to measure the peak perceived torque; this was performed 48 times. They found a tightness of 84%, with only one of these latter screws stripping the bone on insertion (2%).

Two studies used real-time torque feedback via visual displays.[12,15] Mears et al inserted 10 screws into osteoporotic human humeri at 90° or 180° rotation past the point of first screw head contact, to 1.4 Nm or to two-finger tightness.[15] Whilst not explicitly defined, 1.4 Nm was likely chosen as this is 70% of the maximum value for these osteoporotic bones and matches the value found for the optimum tightness in the study by Tankard et al.[18] The torque values that screws were inserted to were recorded, although without direct assessment of the stripping torque. However, based on the assumption that 70% tightness was achieved with the 1.4 Nm tests, tightnesses of 64%, 87% and 70% were generated for the different methods respectively. They found that two out of ten screws were still stripped despite targeting a value of 1.4 Nm, with zero, three and two screws stripping the bone using the other insertion methods respectively. This may be explained by the insertion torque that was targeted, and used as a reference, being beyond the stripping limit of the bone, rather than the technique causing it. However, as no assessment of the maximum torque for the bone samples was performed, this remains unknown. The other study to use visual feedback was performed by Gustafson et al.[12] They asked five senior surgeons and five attending surgeons to insert eight 4.0 mm cancellous screws into polyurethane bone blocks of 0.1 g/cm^3. First, they were asked to insert eight screws to create 'maximum construct stability'. Screws causing stripping were recorded, being 42% of all insertions for this component of the study. Next, digital torque readings were displayed during insertion for the surgeon to use as feedback. With visual feedback, the rate of bone stripping reduced significantly to 15% ($p < 0.001$). Visual feedback was then removed, with the stripping rate returning to a significantly ($p < 0.007$) higher level of 35%, similar to the first part of testing. Awareness of whether the screw holes were stripped was not recorded.

Bone stripping rates

There is very limited research into the rates of bone stripping intraoperatively. A study by Andreassen et al investigated the augmentation of screws if the purchase achieved intraoperatively was determined to be inadequate.[17] In their selected patients (those over 50 years old with isolated ankle fractures), they found that a synthetic bone void filler was needed for 38% of screws, with 88% of patients having at least one screw that required this.[17] The remaining data on rates of stripping come from in vitro studies.[6,9,10,12–15] The range of mean average stripping rates, when reported, was 0–53%[10,15] (Table 1), though some individuals within studies stripped more: up to 83%.[10] Only 19 of the 48 parts of the experiments within studies examining surgical techniques recorded whether screws were stripped, with stripping being confirmed if the torque created after the surgeon had stopped tightening was quantitatively less than the stopping torque. It may be that there was no concern regarding stripping and/or no occurrence of this, explaining why it was not reported. Even when the stripping rates are described, this potentially overlooks the screws inserted beyond the yield torque for the material if recording relies on surgeons' perception. Dinah et al reported 9% (18/200) of samples being stripped inadvertently; however, they also graded unstripped insertions, finding a further 12% (24/200) were deemed to have been overtightened (90–99% of maximum torque).[14]

Few articles have assessed surgeons' subjective abilities to detect whether they had stripped the screw. Gustafson et al showed no correlation between occurrence and the perception of stripping ($p = 0.768$).[12] With visual feedback, there was increased accuracy in predicting stripping, as one would expect if able to watch a digital readout of the applied torque. However, surprisingly, stripping still occurred in 15% of insertions. No data were provided for the rate of accurate predictions when visual feedback was removed, though it was reported as being significantly ($p = 0.008$) better than the 6.1% prediction rate at the start of the experiments. Interestingly, when the visual feedback was removed, whilst the improved perception of stripping was partially maintained, there was no significant reduction in the rate of stripping overall, potentially showing a reliance on augmented feedback. This is also the only study that has investigated any retention of a new method or improvement in technique, though over a very short time period, i.e. within the same experimental setting. Stoesz et al found 45% of screws were stripped on insertion, yet only 10 of 109 (9%) of stripped screws were identified correctly.[10] Identification only occurred when significantly ($p = 0.005$) past the stripping torque (residual torque being 55% of the maximum, compared to 80% when not detected). Whilst not included in this review as the tightness created was not recorded, one study has attempted to quantify practitioners' confidence with the screws they inserted.[19] Siddiqui et al asked one nurse, one junior surgical trainee, one senior surgical trainee and one consultant to insert 4.0 mm partially threaded cancellous screws into chipboard and asked them to assign each insertion (n = 30, 43, 35 and 34 respectively) a score 0–10; with 0 being very weak and 10 being very strong. Each screw then underwent axial pullout. They found correlations between axial pullout force and confidence scoring of r^2 = 0.34, 0.26, 0.22 and 0.45 respectively. Unfortunately, the material used, the lack of torsional force assessment or whether samples were stripped on insertion, and the variability in the material properties between different chipboard samples greatly limits the generalizability of these findings to clinical practice.

The effect of surgical experience on screw tightness and stripping rates

Stripping rates appear to be individual to each surgeon, with a wide range of performance. There is both intra- and inter-surgeon variation in insertional torque.[9] A stark example of variations between surgeons is seen in Gustafson et al where, when inserting screws without visual feedback, one of their 10 surgeons stripped 16 out of 16 and another 15 out of 16 samples.[12] Conversely, of their 10 surgeons, two stripped none, and a further two only stripped one sample when there was no visual feedback.

The torsional force applied to screws increases with more surgical experience,[7,9,16] but this also increases the rate of stripping.[9] Wilkofsky et al found that more experienced surgeons applied significantly more torque to screws than either first and second-year ($p = 0.003$) or third and fourth-year surgical trainees ($p = 0.007$).[16] This resulted in a greater tightness of 83±12% compared to 70±19% and 71±20% respectively. Whilst the variation in tightness was less for the most experienced surgeons than other groups, the lateral motion generated whilst creating the higher insertional forces, i.e. non-coaxial insertion or 'screw wobble', was approximately 56% larger ($p < 0.05$). This study also found that the number of screw rotations varied greatly between surgeons, ranging from four to 21 revolutions being needed when inserting a 3.5 mm cortical screw into polyurethane bone.[16] Apparently contradicting these findings, as previously stated, Acker et al found no significant difference in the applied torque between first (junior) and fifth-year (senior) trainees, nor when compared to senior surgeons.[9] Generally, concerns during screw insertion are related to the balancing of the appropriate minimum tightness for the construct to generate sufficient purchase and resistance to failure during locomotion, against overtightening the screw and causing bone stripping. However, more trainee experience appears to lead to an increased chance of this more detrimental, latter situation occurring.[9] As no optimum tightness was established for the bone model used in their study (as a function of compression and pullout strength) it is unknown how tight screws should have been inserted, just that stripping the bone should have been avoided. Stoesz et al found no relationship between stripping rates and surgeon experience ($p = 0.437$), but in a comparison of ten surgeons, there were significant differences in stripping rates between individuals ($p < 0.001$);[10] the percentage of samples stripped ranged from 17% to 83%. Seven of the 131 unstripped screws were thought by surgeons to be stripped, however, six of these reports were from one surgeon. These two aspects strongly justify including multiple surgeons in any study investigating techniques in order to reduce the impact from different abilities, or at least ensuring that potential variations between surgeons are taken into account and reported.

Effect of different instructions to surgeons on screw tightness and stripping rates

With the exception of one article that compared four different instructions[15] and the in vivo study,[17] the instructions given to surgeons during these studies fall into five categories: subjective feeling of tightness,[8,11] 'two-fingers tight',[7,9,14,16] optimal for good fixation,[6] maximum construct stability[10,12] or maximum holding force without stripping the bone.[13] There are no direct comparisons between different instructions to surgeons to know whether any of these methods make a difference to the techniques employed. 'Two-fingers tight' has been reported to be the gold standard for screw insertion, if performed by an experienced orthopaedic surgeon[15], and is commonly taught in theatres to trainee surgeons in the USA and Europe.[9,20] However, the evidence that this technique improves screw insertion is limited, and subsequently, when evaluated, has been shown to be incorrect in that it does not lead to a consistent level of torque being applied.[9,21] Further to this, previous work, such as Cordey et al, has been reported by others to have used two-finger tightness,[15,18] despite not defining this in their methods.[6] Targeting a specific tightness and comparing surgeons' ability to repeatably and accurately achieve this target versus other tightnesses has not been investigated, nor has the effect of different instructions on the same physical variables.

Effect of variations in bone density on screw tightness and stripping rates

A common issue with biomechanical research is the model used for testing. It is established that artificial bone models reduce variability, costs and ethical concerns. However, they do not demonstrate many of the biomechanical characteristics of real bone, such as cortical porosity and failure mechanisms, thus limiting generalizability from any research using them. In contrast, in vitro cadaveric human and animal bone models will generate more realistic resistances to screw insertion, but the variability in some models, even between contralateral pairs,[22] means that appropriately powered results can be difficult to generate. Animal models[23] can address these issues, but ultimately, unless in vivo human bone is used, results may not be fully translatable into clinical practice.

Nicayenzi et al showed no significant difference between human and artificial femora cortices when inserting cortical screws in terms of the stripping torque when normalized to adjust for changes in bone geometries, measured by the bone–screw interface area: normalized stripping torque = stripping torque/($\pi \cdot$ screw major diameter $\cdot$ cortical thickness).[24] When comparing maximum torque to the plateau torque during insertion, and comparing these as predictive variables, Reynolds et al found no difference between ovine and human bone maximum torques ($p = 0.331$). Despite also using synthetic bone blocks, they did not report whether there were significant differences in comparison to these.[25]

Aziz et al compared fresh frozen, embalmed and dried adult human humeri alongside normal and low density synthetic bones.[8] They found that when one surgeon inserted cortical screws to ‘a subjective feeling of tightness’, there was no difference in the tightness between any of the models used. With cancellous screws, one difference was detected with the tightness being 13% lower ($p < 0.05$) in fresh-frozen bone than in artificial osteoporotic bone, though the authors reported that all comparisons were underpowered.

Tsuji et al showed, for cortical screws in synthetic bone, as density increased from 0.08 to 0.80 g/cm^3, screw tightness decreased ($R = -0.63$), yet in human cadaveric femora, there was no difference in terms of density changes.[11] For cancellous screws in artificial bone models, they found as density increased from 0.08 to 0.64 g/cm^3, screw tightness increased ($R = 0.59$), yet in human cadaveric femora, the opposite was seen ($R = -0.56$). This shows the potential variability in the insertion technique of the same individual, given that with different screws in the same material or the same screw in different material, different trends were seen each time. All other studies involving human bone did not have any comparison with artificial models.[6,7,13–15,17] Studies just using artificial bones have shown neither an effect on achieved tightness ($p = 0.299$) nor on stripping rates due to bone density ($p = 0.186$).[10]

Unassessed variables

Numerous variables related to the practical insertion of screws have not been investigated. No studies have compared different sizes of the same style screw (i.e. cortical screws with outer screw thread diameters of 3.5 mm and 4.5 mm) within the same group of participants or bone models. This would provide information on the ability of surgeons to adapt to different commonly used screw sizes and whether different size screws are more prone to stripping. Studies directly comparing cortical and cancellous screw insertion techniques are limited as either the number of cortices engaged is different[14] and/or the outer diameter of the screw is considerably greater for cancellous screws.[8,11] No tests on the effect of cortical thickness on the screw tightness generated by surgeons have been performed, which could highlight situations where extra care was required to prevent bone stripping.

No analyses of the contributions of cancellous and cortical bone have been performed to elucidate which aspects contribute most to the proprioceptive feedback experienced by a surgeon in the presence of both classes of bone. Cancellous bone density and microarchitecture have been shown to affect insertion failure torques[26] though not the tightness applied by surgeons. Focussing on screw insertion into cancellous bone may be important, given that the density of this bone is less than that of cortical, and thus stripping rates are higher. Some studies have focused on a pure cancellous model, i.e. no cortical shell present.[10] Whilst this highlights a situation where bone damage may be easier to cause, all clinically inserted non-locking screws are likely to have a near cortex of bone, and if this is greater than 1.5 mm, the role of the cancellous bone has been shown to be limited.[27]

Other practical surgical variables are yet to be investigated. Whilst it is expected that gloves were worn by surgeons during experiments, and certainly during in vivo testing,[17] no studies explicitly stated their use, despite it being unknown whether different types of gloves (such as unsterile compared to sterile), or number of layers (single compared to double layer) affect screw tightness. The effect of screw and screwdriver variables may additionally impact on screw tightness; aspects such as screw head shape, handle shape and presence or absence of a washer or plate.

Limitations

There are limitations with this review. First, some assumptions have been made when analysing the data provided by authors, such as assuming that 1.4 Nm was chosen by Mears et al[15] as this reflects 70% of Tmax for osteoporotic samples in the study by Tankard et al.[18] Second, there is only one in vivo study reported, likely due to the difficulties with measuring or predicting the stripping torque intraoperatively without causing additional damage to the bone. Third, no screws with an outer diameter smaller than 3.5 mm have been assessed, nor have screws of different shapes such as those with conical inner and outer diameters. There may be limited generalizability of the findings of these studies to other situations. Finally, what values have been used to calculate tightness are unclear in some studies. If the torque applied during insertion is greater than the torque that

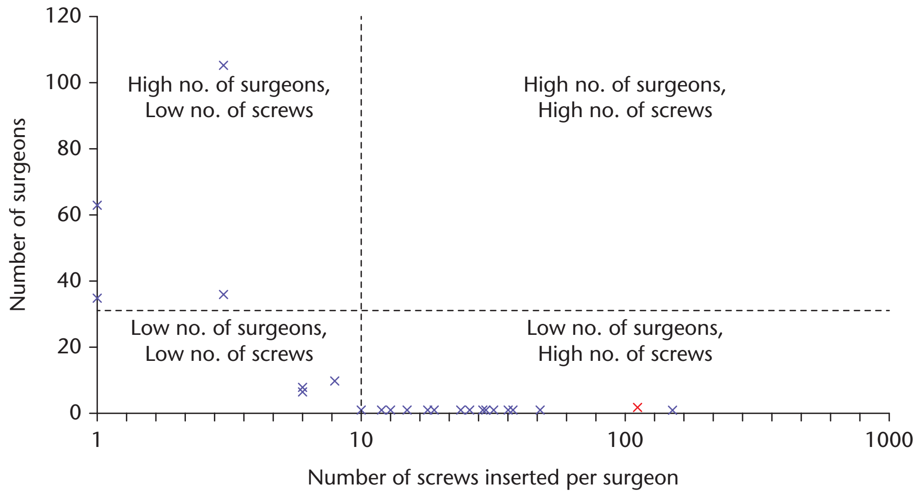

Fig. 3 Number of surgeons and the number of screws each inserted within each part of each study (12 studies reviewed in this manuscript, 48 experiments in total), with the latter displayed logarithmically. Blue markers indicate in vitro studies, red marker for the sole in vivo study.[17] High/low surgeon and screw number quadrants created based on more or less than 30 surgeons and more or less than 10 screws inserted by each.

can subsequently be applied when attempting to strip the surrounding material, when converting this to a percentage of the latter, values greater than 100% will be calculated. If these are included in the mean average, it will skew results, such as the fifth-year residents in the study by Acker et al reporting a mean of 103%. Future studies can eliminate this skewing factor by reporting the unstripped tightness and stripping rates.

When summating all the screw insertions within the 12 different studies in this review article, a total of 260 surgeons were involved. These surgeons inserted a total of 2793 screws, an average of 11 screws each, although only 1510 screws have been inserted by 145 surgeons where the tightness was measured. The maximum number of screws inserted under the same in vitro conditions with one surgeon is 160 but only eight if more than two surgeons were used (Fig. 3). Thus, an average is used to describe the behaviours of the entire orthopaedic, and potentially wider, communities such as maxillofacial, plastic and neurosurgery. Further to this, the optimum tightness for primary fixation is currently unknown (except for it being below the stripping torque). Additionally, the effect that screw tightness has on fracture healing remains unknown and would require complicated in vivo studies given the multifactorial nature of bone remodelling. In vivo tightness and its effects on healing may be very different to the biomechanical experiments performed on non-living tissues. Thus, all assessments of techniques are limited as even if repeatable screw tightness is achievable, which appears not to be the case, these values for tightness may not be what is required to generate optimal fixation.

Conclusions

Considerable variation has been found in the tightness applied to screws by surgeons, with combined averages of 78±10% for cortical screws and 80±6% for cancellous screws across all studies. When specifically investigated, nearly 26% of all screws have been found to have stripped the surrounding bone during insertion, with most of these occurrences being undetected by the surgeon. Large variation between different surgeons has been found, with some studies finding contradictory outcomes regarding whether more surgical experience is associated with improvements in tightness or rate of stripping. Whilst some variables have been investigated for their impact on screw tightness, such as the effect of different bone densities, many remain unexplored. Future work to establish the influence from different intraoperative situations such as cortical thickness and screw diameter could highlight areas where extra vigilance is needed to avoid overtightening of screws, whilst all future studies should include multiple surgeons to reduce the risk of individual surgeon biases. Further research establishing the optimum tightness for screw constructs is needed to help surgeons by providing a target torque for each screw, alongside integration of automating torque detection during screw insertion to prevent excess damage being caused. The development of augmented screwdrivers able to indicate the optimum and maximum torques for any given situation would greatly help with surgical education and clinical performance. This could make screw insertion more efficient through higher quality screw fixation generating more secure constructs, reducing fixation failure rates.

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